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Development and validation of a finite-element musculoskeletal model incorporating a deformable contact model of the hip joint during gait

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**Key words:** Musculoskeletal modelling; finite element; contact mechanics; articular cartilage; hip joint; gait

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## Abstract

Musculoskeletal models provide non-invasive and subject-specific biomechanical investigations of the musculoskeletal system. In a musculoskeletal model, muscle forces contribute to the deformation and kinematics of the joint which in turn would alter moment arms of muscles and ground reaction forces and thus affect the prediction of muscle forces and contact forces and contact mechanics of the joint. By far, deformable contact models of the hip have not been considered in musculoskeletal models, and the role of kinematics and deformation within the hip in muscle forces and hip contact mechanics is unknown. In this study, an FE musculoskeletal model including bones, joints and muscles of the lower extremity was developed. A deformable contact model of the hip joint was incorporated and coupled into the musculoskeletal model. Joint angles and ground reaction forces during gait were used as inputs. Optimization minimizing the sum of muscle stresses squared was performed directly to the FE musculoskeletal model in order to simultaneously solve muscle forces and contact forces and contact stresses of the hip joint within a single framework. The calculated hip contact forces corresponded well to the *in vivo* measurement data. The maximum hip contact stress was 6.5 MPa and occurred at weight-acceptance. The influence of kinematics and deformation in the hip on muscles forces and hip contact forces was minimal and not sensitive to variations in the thickness and properties of the joint cartilage during gait. This suggests that the uncoupled approach in which the hip contact forces and contact mechanics are simulated in separate frameworks would serve as an effective and efficient alternative for subject-specific modelling of the hip. This study provides guidance for the level of complexity needed for future hip models and can be used to evaluate biomechanical changes of the musculoskeletal system following interventions.

## 1. Introduction

Numerical analyses allow for non-invasive and systematic biomechanical evaluation of the hip joint. For example, finite-element (FE) models have been widely used to study stresses and strains within the hip (Henak et al., 2014, Li et al., 2016). These regional hip models require contact forces of the joint as inputs. Therefore, musculoskeletal models of the lower extremity bridging the body kinematics and joint biomechanics are needed for non-invasive and subject-specific studies.

Previous musculoskeletal models (e.g. OpenSim (Delp et al., 2007) and AnyBody (AnyBody Technology, Denmark)) have been mostly constructed in multi-body dynamics with the model assumed as rigid and the hip as a simple three degrees-of-freedom (DOFs) rotational joint (Li et al., 2015). Based on kinematics and kinetics of the lower extremity, these multi-body dynamics models have been primarily used to determine contact forces of the joint which then serve as inputs for FE models of the joint region to simulate its contact mechanics (i.e. uncoupled simulation) (Farrokhi et al., 2011). Contact joint models have been recently incorporated into multi-body dynamics musculoskeletal models by calculating contact forces between rigid bodies based on their overlapping volume (Zhang et al., 2015). However, in this approach, the joint kinematics contributed by the cartilage deformation cannot be realistically simulated.

In a musculoskeletal model, muscle forces contribute to the deformation and kinematics of the joint which in turn would alter moment arms of muscles and ground reaction forces and thus affect the prediction of muscle forces and contact forces and contact mechanics of the joint. This interaction was found to have a marked effect on the biomechanics of the knee through some recent attempts in which the knee joint was presented in detail and coupled into a musculoskeletal model (Marouane et al., 2017, Shu et al., 2018, Hume et al., 2019), but remains unknown for the hip. In order to consider the interaction between muscle forces and kinematics and contact mechanics in the joint, incorporation of deformable and contact joint models into musculoskeletal models is needed. However, such models involve more complex construction and optimization procedures and longer simulation periods (Shu et al., 2018, Hume et al., 2019), compared with the uncoupled approach simulated in separate frameworks. So far, deformable contact models of the hip have not been considered in musculoskeletal models, and the role of kinematics and deformation within the hip in muscle forces and hip contact mechanics is poorly understood. This information provides important guidance for the level of complexity needed for future musculoskeletal models focused on the hip joint, so that a reasonable balance between accuracy and efficiency for numerical simulations of the hip joint can be determined.

The aim of this study was to develop and validate an FE musculoskeletal model of the lower extremity incorporating a deformable contact model of the hip joint. Additionally, the effect of kinematics within the hip on muscle forces and hip contact mechanics was evaluated by comparing the predictions of this coupled model to an uncoupled model in which contact forces and contact stresses of the hip were simulated using separate frameworks.

## 2. Methods

### 2.1. Musculoskeletal model

The FE musculoskeletal model was developed in an implicit FE solver FEBio (version 2.6.4; <http://febio.org/febio>). The model included the bones and joints of the right lower extremity and the complete set of muscles driving the hip which were modelled as contractile forces (**Fig. 1a**) (Carbone et al., 2015). The insertion and origin of the muscles were revised based on the refined TLEM 2.0 model (De Pieri et al., 2018). To ensure proper computational efficiency, the muscles that do not cross the hip joint were excluded; the knee and ankle were assumed as simple three DOFs rotational joints and bones as rigid, with the patella immobilized onto the tibia (Li et al., 2015).

A natural hip model from a 55 year-old, 109 kg, 180 cm male was incorporated into the FE musculoskeletal model, considering the cartilage with subject-specific geometry (Li et al., 2016). The modelling of the hip joint is based on a previously validated procedure (Li et al., 2014). The back surfaces of the cartilage were bonded onto the underlying bones. Frictionless contact between the cartilage layers was defined, with the surface of the femoral head cartilage as the master and the surface of the acetabular cartilage as the slave. The cartilage and bones were represented by 11460 eight-noded hexahedral elements and 6145 four-noded tetrahedral elements, respectively. The mesh density of the cartilage was evaluated to ensure that the differences in the peak contact stress were below 5% when the number of elements was doubled. Neo-Hookean material was adopted as the baseline constitutive model of the cartilage, with strain energy given by:

$$W = \frac{1}{2}\mu(\tilde{I}_1 - 3) + \frac{1}{2}K(\ln(J))^2 \quad (1)$$

Where,  $\tilde{I}_1$  is the first deviatoric invariant of the right Cauchy deformation tensor;  $J$  Jacobian of the deformation;  $\mu$  shear modulus;  $K$  bulk modulus. The cartilage material was reinforced by fibres with isotropic distribution. The fibre strain energy is given by:

$$W_f = \int_A H(\tilde{I}_n - 1) \xi(\mathbf{n}) (\tilde{I}_n - 1)^{\beta(\mathbf{n})} dA \quad (2)$$

Where,  $\tilde{I}_n$  is the square of the deviatoric fibre stretch;  $\mathbf{n}$  the unit vector along the fibre direction; The integral is evaluated over the unit sphere  $A$  spanned by all directions  $\mathbf{n}$ ;  $H(-)$  the unit step function ensuring that fibres only sustain tension;  $\xi$  scales the fibre response;  $\beta$  controls the nonlinearity of fibres. Refer to (Maas and Weiss, 2007) for further description of the constitutive model. Coefficients of the constitutive model were defined as:  $\mu = 1.82$  MPa,  $K = 1860$  MPa,  $\xi = 9.19$  MPa, and  $\beta = 4$  (Henak et al., 2014).

### 2.2. Subject and gait data

Inputs of the FE musculoskeletal model including joint angles and ground reaction forces are based on the gait data of a patient (named as H2R in the database; a 62 year-old, 78 kg, 172 cm male) with an instrumented hip implant during walking (<https://orthoload.com/>) (Bergmann et al., 2016). The FE musculoskeletal model was linearly scaled to match the anthropometry of the patient's lower extremity. As the

simulation was quasi-static and the inertia effect not considered, the pelvis was immobilized along the three translational DOFs. The rotational angles of the pelvis (relative to the global coordinate system), hip, knee and ankle of the patient were derived from the marker trajectories in the gait data using Visual 3D (V6; C-Motion, USA). These angles were then used to rotate the pelvis, hip, knee and ankle in the musculoskeletal model. The ground reaction forces and moments along the axes of the global coordinate system were distributed onto the heel and toe of the musculoskeletal model according to the locomotion of the ground reaction forces on the force plates, so that these forces and moments were applied onto the proper position of the foot.

### 2.3. Calculation of muscle forces and hip contact mechanics

An optimization approach was developed in this study to solve the muscle redundancy issue in the musculoskeletal system (i.e. muscles outnumber the equations of equilibrium, requiring optimization to determine a unique solution of muscle forces). Based on the muscle forces and the corresponding joint moments in the FE musculoskeletal model, the muscle forces were optimized until the sum of the square of muscle stresses (i.e. muscle force over physiological cross-sectional area) was minimized, and at the same time, the resultant hip moment approached zero. The “fmincon” optimization tool in MATLAB (R2017a, Mathworks, MA) was adopted to solve the optimization problem. Both the FE simulation and the optimization were continuous, e.g. simulation of the FE musculoskeletal model at 0.5s starting from the optimized model at its previous time instance (i.e. 0.45s). The outputs of the model including muscle forces and contact forces, contact stresses and kinematics of the hip were analysed at 14 evenly distributed time instances of the stance phase of a gait, starting from heel-strike (0s, 0%) to toe-off (0.65s, 100%). Kinematics of the hip was calculated as the translational displacement of the femoral head center relative to the acetabulum center.

In order to validate the model, the hip contact forces predicted by the FE musculoskeletal model were compared to the *in vivo* measurement data during the same gait trial. Additionally, a musculoskeletal model with three DOFs rotational hip joint was developed and its predicted contact forces and kinematics of the hip were then used as the inputs of a regional FE model of the same hip joint (**Fig. 1b**). Results of this uncouple simulation (i.e. uncoupled model) were compared to the original musculoskeletal model incorporating a contact hip joint (i.e. coupled model), in order to evaluate the effect of kinematics and deformation of the hip joint on its contact mechanics and muscle forces.

In this study, only one subject was investigated. However, different subjects have various geometry and properties of the hip cartilage, which might lead to altered conclusion regarding the influence of hip kinematics and deformation on hip contact mechanics and muscle forces. To account for the variations in the thickness and properties of the joint cartilage, a sensitivity study was conducted by performing the comparison (i.e. coupled approach VS uncoupled approach) using four other models that were constructed based on the original cartilage model: Model 2 with approximately 25% thicker cartilage (Shepherd and Seedhom, 1999); Model 3 with approximately 25% thinner cartilage; Model 4 with 50% higher  $\mu$  and  $\xi$ ; Model 5 with 50% lower  $\mu$  and  $\xi$

(Athanasίου et al., 1994). Models 3 and 4 were developed by removing/adding a layer of elements based on the original model.

## Results

As shown in **Fig. 2**, predictions of the FE musculoskeletal model, including the direction and magnitude of the hip contact forces and the timing at heel-strike, weight-acceptance, mid-stance, push-off and toe-off, were in good agreement with the *in vivo* measurement data. Compared to the *in vivo* measurement data, the simulated total hip contact force (i.e. sum of the three vector components in **Fig. 2**) was 25% higher at weight-acceptance, 7% lower at mid-stance and 2% higher at push-off, with a mean absolute percentage error of 15% over the stance cycle.

The difference in the hip contact forces between the coupled (original) and uncoupled models was less than 1% (**Fig. 2**). The magnitude and distribution of the hip contact stresses between the coupled (original) and uncoupled models were also nearly identical (**Fig. 3**). Contact stresses at weight-acceptance and push-off were markedly higher than the other gait phases, with the maximum value of 6.5 MPa that occurred at weight-acceptance (**Fig. 3**). The difference in the forces of the major hip muscles between the coupled (original) and uncoupled models was within 5% (**Fig. 4**). As predicted by the coupled (original) model, kinematics and deformation in the hip occurred during walking was less than 1 mm which was minimal compared to the scale of the joint (**Fig. 5**). In the sensitivity study (Models 2-5), the differences in hip contact forces and muscles forces between the coupled and uncoupled approaches were within 5%.

## Discussion

In this study, an FE musculoskeletal model of the lower extremity incorporating a contact model of the hip joint was developed for the first time. Optimization was performed directly to the FE musculoskeletal model in order to simultaneously solve the muscle forces and the contact forces and contact stresses of the hip joint within a single framework. The hip contact forces predicted by the model corresponded well to the *in vivo* measurement data over the entire stance cycle. The maximum contact stress in the hip during walking predicted by the model was 6.5 MPa under a load of 1982 N which is consistent with previous *in vitro* measurements (4–10 MPa under loads of 2500 N–3000 N) (Brown and Shaw, 1983, Afoke et al., 1987, Anderson et al., 2008).

FE musculoskeletal models incorporating deformable contact joints enable simulations of the interaction between muscle forces and joint kinematics/deformation which cannot be accounted for in the widely used multi-body dynamics musculoskeletal models, allowing for more systematic and realistic biomechanical analyses of the musculoskeletal system, but at the same time involving more complex and lengthy simulations

141 (Shu et al., 2018, Hume et al., 2019, Sharifi et al., 2020). Using 8 CPU cores, the simulations of the coupled  
142 model and uncoupled model required 3 days and 3.5 hours, respectively. In this study, it was found that  
143 kinematics and deformation in the natural hip joint was markedly smaller than the dimension of the hip and  
144 the moment arms of muscles and ground reaction forces, and thus had a minimal effect on the muscles forces  
145 and hip contact forces calculated through musculoskeletal models. This suggests that the uncoupled approach  
146 in which the loading and contact mechanics of the hip are simulated in separate frameworks would serve as an  
147 effective and efficient alternative for subject-specific modelling of the hip. This finding is further supported  
148 by the sensitivity study in which it was found that variations in the geometry and material properties of the hip  
149 cartilage had a minimal effect on the hip contact forces and muscle forces. As a healthy hip joint during walking  
150 was evaluated in this study, further analyses should be performed for other activities and for hips in dysplasia  
151 in which the joint is less congruent than a healthy hip and its kinematics might have an evident influence on  
152 the muscle forces and hip contact mechanics (Lequesne et al., 2004).

153       There are some limitations. First, the optimization was performed only for the hip joint. Including  
154 multiple joints in the optimization requires extra muscles and increased computational expenses, but would  
155 enable more realistic modelling. As found by Adouni and Shirazi-Adl (2013), inclusion of the hip joint in the  
156 optimization of the knee and ankle joints slightly influences the calculation of muscle forces and contact forces  
157 of the knee. Furthermore, consideration of the muscles across the other joints (e.g. knee) would also improve  
158 the modelling accuracy, as these muscles might affect the calculation of hip muscle forces and the resultant  
159 joint contact forces. Another limitation is that parameters of the muscle including its passive properties, 3D  
160 geometry, large attachment areas, spatial fibre alignment within muscles, and contact and wrapping between  
161 muscles and surrounding tissues are important for the accuracy of musculoskeletal modelling but were not  
162 considered in this study. These aspects can be accounted for by incorporating 3D muscles into musculoskeletal  
163 models which require a lengthy period of simulations (Li et al., 2019). Additionally, ligaments and capsules  
164 were not accounted for, because these soft tissues were found to contribute slightly to the kinematics and  
165 contact mechanics of the hip in a musculoskeletal model representing these tissues as a 1D spring (Zhang et  
166 al., 2015). However, the role of ligaments and capsules in the hip should be further assessed using detailed 3D  
167 models. Future development will focus on creating a hybrid musculoskeletal model incorporating both 1D and  
168 3D muscles and other soft tissues as a reasonable compromise between accuracy and efficiency. The labrum  
169 was excluded because it provides little assistance in load bearing of the hip (Henak et al., 2011). The time-  
170 dependent biphasic/viscoelastic properties of the cartilage was not considered, because it is highly time-  
171 consuming to achieve numerical convergence in biphasic simulations and the time-dependent response of the  
172 hip cartilage is minimal during short term loading (Li et al., 2013, Li et al., 2016, Todd et al., 2018).

173       Generally, the hip contact forces predicted by the computer model corresponded reasonably well to the  
174 *in vivo* measurement over the entire stance cycle. The difference in comparison might be due to several reasons,  
175 apart from the model simplification described in the paragraph above and errors of the *in vivo* measurement.  
176 First, the boundary conditions of the experimental models were derived from the gait data of a patient with an  
177 instrumented hip implant for the purpose of validation, whereas the musculoskeletal model and geometric



178 model of the finite element hip joint were from subjects with healthy hip joints. Validation and *in vivo*  
179 measurement of biomechanics in healthy hip joints are challenging, but should be attempted, e.g., through  
180 imaging measurements and validation of soft tissue deformation and joint kinematics using the same subject.  
181 Secondly, muscle forces predicted by the models were not compared to the experimental data such as  
182 electromyography (EMG) signals, because of uncertainties in acquisition and conversion of EMG signals.  
183 Although validation of muscle forces was not within the scope of the current study, inclusion of experimentally  
184 measured muscle activity either in the validation or among the model inputs would contribute to the accuracy  
185 and validity of future models focusing on biomechanics of muscles.

186 In this study, a musculoskeletal model with a contact joint was developed within a single finite element  
187 framework, with optimization integrated into the finite element simulation process. This enables simulations  
188 of 3D geometries, deformation and biotribology of joints, bones, muscles (Li et al., 2019) and other tissues  
189 within musculoskeletal models. The modelling framework also allows for multi-scale analyses, considering  
190 interactions between models at different scales spanning from the skeletal scale to the tissue and micro scales.  
191 Additionally, the modelling framework can be used to evaluate biomechanical changes of the musculoskeletal  
192 system following interventions.

193

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197

#### 198 **Conflict of interest**

199 The author declares no conflict of personal or financial interests.

## 200    **References**

- 201    ADOUNI, M. & SHIRAZI-ADL, A. 2013. Consideration of equilibrium equations at the hip joint alongside those  
202                    at the knee and ankle joints has mixed effects on knee joint response during gait. *Journal of*  
203                    *Biomechanics*, 46, 619-624.
- 204    AFOKE, N. Y., BYERS, P. D. & HUTTON, W. C. 1987. Contact pressures in the human hip joint. *Journal of Bone*  
205                    *and Joint Surgery (British)*, 69, 536-41.
- 206    ANDERSON, A. E., ELLIS, B. J., MAAS, S. A., PETERS, C. L. & WEISS, J. A. 2008. Validation of finite element  
207                    predictions of cartilage contact pressure in the human hip joint. *Journal of Biomechanical Engineering*,  
208                    130, 051008.
- 209    ATHANASIOU, K. A., AGARWAL, A. & DZIDA, F. J. 1994. Comparative study of the intrinsic mechanical  
210                    properties of the human acetabular and femoral head cartilage. *Journal of Orthopaedic Research*, 12,  
211                    340-349.
- 212    BERGMANN, G., BENDER, A., DYMKE, J., DUDA, G. & DAMM, P. 2016. Standardized loads acting in hip  
213                    implants. *PLOS ONE*, 11, e0155612.
- 214    BROWN, T. D. & SHAW, D. T. 1983. In vitro contact stress distributions in the natural human hip. *Journal of*  
215                    *Biomechanics*, 16, 373-384.
- 216    CARBONE, V., FLUIT, R., PELLIKAAN, P., VAN DER KROGT, M. M., JANSSEN, D., DAMSGAARD, M., VIGNERON,  
217                    L., FEILKAS, T., KOOPMAN, H. F. J. M. & VERDONSCHOT, N. 2015. TLEM 2.0 – A comprehensive  
218                    musculoskeletal geometry dataset for subject-specific modeling of lower extremity. *Journal of*  
219                    *Biomechanics*, 48, 734-741.
- 220    DE PIERI, E., LUND, M. E., GOPALAKRISHNAN, A., RASMUSSEN, K. P., LUNN, D. E. & FERGUSON, S. J. 2018.  
221                    Refining muscle geometry and wrapping in the TLEM 2 model for improved hip contact force  
222                    prediction. *PLOS ONE*, 13, e0204109.
- 223    DELP, S. L., ANDERSON, F. C., ARNOLD, A. S., LOAN, P., HABIB, A., JOHN, C. T., GUENDELMAN, E. & THELEN,  
224                    D. G. 2007. OpenSim: Open-Source Software to Create and Analyze Dynamic Simulations of  
225                    Movement. *Biomedical Engineering, IEEE Transactions on*, 54, 1940-1950.
- 226    FARROKHI, S., KEYAK, J. H. & POWERS, C. M. 2011. Individuals with patellofemoral pain exhibit greater  
227                    patellofemoral joint stress: a finite element analysis study. *Osteoarthritis and Cartilage*, 19, 287-294.
- 228    HENAK, C. R., ATESHIAN, G. A. & WEISS, J. A. 2014. Finite element prediction of transchondral stress and  
229                    strain in the human hip. *Journal of Biomechanical Engineering*, 136, 021021-021021.
- 230    HENAK, C. R., ELLIS, B. J., HARRIS, M. D., ANDERSON, A. E., PETERS, C. L. & WEISS, J. A. 2011. Role of the  
231                    acetabular labrum in load support across the hip joint. *Journal of Biomechanics*, 44, 2201-2206.
- 232    HUME, D. R., NAVACCHIA, A., RULLKOETTER, P. J. & SHELBURNE, K. B. 2019. A lower extremity model for  
233                    muscle-driven simulation of activity using explicit finite element modeling. *Journal of Biomechanics*,  
234                    84, 153-160.
- 235    LEQUESNE, M., MALGHEM, J. & DION, E. 2004. The normal hip joint space: variations in width, shape, and  
236                    architecture on 223 pelvic radiographs. *Annals of the Rheumatic Diseases*, 63, 1145.
- 237    LI, J., HUA, X., JONES, A. C., WILLIAMS, S., JIN, Z., FISHER, J. & WILCOX, R. K. 2016. The influence of the  
238                    representation of collagen fibre organisation on the cartilage contact mechanics of the hip joint.  
239                    *Journal of Biomechanics*, 49, 1679-1685.
- 240    LI, J., LU, Y., MILLER, S., JIN, Z. & HUA, X. 2019. Development of a finite element musculoskeletal model with  
241                    the ability to predict contractions of three-dimensional muscles. *Journal of Biomechanics*, 94, 230-  
242                    234.
- 243    LI, J., MCWILLIAMS, A. B., JIN, Z., FISHER, J., STONE, M. H., REDMOND, A. C. & STEWART, T. D. 2015. Unilateral  
244                    total hip replacement patients with symptomatic leg length inequality have abnormal hip  
245                    biomechanics during walking. *Clinical Biomechanics* 30, 513-519.
- 246    LI, J., STEWART, T. D., JIN, Z., WILCOX, R. K. & FISHER, J. 2013. The influence of size, clearance, cartilage  
247                    properties, thickness and hemiarthroplasty on the contact mechanics of the hip joint with biphasic  
248                    layers. *Journal of Biomechanics*, 46, 1641-1647.
- 249    LI, J., WANG, Q., JIN, Z., WILLIAMS, S., FISHER, J. & WILCOX, R. K. 2014. Experimental validation of a new  
250                    biphasic model of the contact mechanics of the porcine hip. *Proceedings of the Institution of*  
251                    *Mechanical Engineers, Part H: Journal of Engineering in Medicine*, 228, 547-555.

252 MAAS, S. A. & WEISS, J. A. 2007. *FEBio Theory Manual*, <http://mrl.sci.utah.edu/software/febio>.

253 MAROUANE, H., SHIRAZI-ADL, A. & ADOUNI, M. 2017. 3D active-passive response of human knee joint in gait  
 254 is markedly altered when simulated as a planar 2D joint. *Biomechanics and Modeling in*  
 255 *Mechanobiology*, 16, 693-703.

256 SHARIFI, M., SHIRAZI-ADL, A. & MAROUANE, H. 2020. Sensitivity of the knee joint response, muscle forces  
 257 and stability to variations in gait kinematics-kinetics. *Journal of Biomechanics*, 99, 109472.

258 SHEPHERD, D. E. & SEEDHOM, B. B. 1999. Thickness of human articular cartilage in joints of the lower limb.  
 259 *Annals of the Rheumatic Diseases*, 58, 27-34.

260 SHU, L., YAMAMOTO, K., YAO, J., SARASWAT, P., LIU, Y., MITSUISHI, M. & SUGITA, N. 2018. A subject-specific  
 261 finite element musculoskeletal framework for mechanics analysis of a total knee replacement.  
 262 *Journal of Biomechanics*, 77, 146-154.

263 TODD, J. N., MAAK, T. G., ATESHIAN, G. A., MAAS, S. A. & WEISS, J. A. 2018. Hip chondrolabral mechanics  
 264 during activities of daily living: Role of the labrum and interstitial fluid pressurization. *Journal of*  
 265 *Biomechanics*, 69, 113-120.

266 ZHANG, X., CHEN, Z., WANG, L., YANG, W., LI, D. & JIN, Z. 2015. Prediction of hip joint load and translation  
 267 using musculoskeletal modelling with force-dependent kinematics and experimental validation.  
 268 *Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine*,  
 269 229, 477-490.

270

## Figure Legends

**Fig. 1.** a – coupled model in which a contact model of the natural hip joint was incorporated into an FE musculoskeletal model of the lower extremity (cartilage displayed in yellow). b – uncoupled model in which the hip contact forces calculated in a musculoskeletal model with three DOFs rotational hip joint were used as the inputs for an FE model of the hip region. The musculoskeletal models include 33 unique hip muscles comprised of 97 musculotendon fibres (displayed in red).

**Fig. 2.** Contact forces in the hip predicted by the musculoskeletal models in comparison with the *in vivo* measurement data (Bergmann et al., 2016). Force components along the anterior-posterior (AP), superior-inferior (SI) and lateral-medial (LM) directions in the lab/global coordinate system are illustrated. The simulated hip contact forces corresponded well to the *in vivo* measurement data. The hip contact forces between the coupled (original) and uncoupled models were nearly identical. In the sensitivity study, hip contact forces of the models with varying thickness and material properties of the cartilage (Models 2-5) were approximately overlapped with the current plots (original model).

**Fig. 3.** Contour of contact stress on the surface of acetabular cartilage at characteristic gait phases predicted by the coupled (original) model, in comparison to the uncoupled model. The magnitude and distribution of the hip contact stresses between these two models were nearly identical. Contact stresses at weight-acceptance and push-off were markedly higher than the other gait phases, with the maximum value of 6.5 MPa that occurred at weight-acceptance.

**Fig. 4.** Forces of the major hip muscles predicted by the coupled (original) model, in comparison to the uncoupled model. The difference in the muscles forces between these two models was less than 5%. In the sensitivity study, muscle forces of the models with varying thickness and material properties of the cartilage (Models 2-5) were approximately overlapped with the current plots (original model).

**Fig. 5.** Kinematics of the hip in the coupled (original) model, calculated as the translational displacement of the femoral head center relative to the acetabulum center. Kinematics of the hip occurred during walking was less than 1 mm and was minimal compared to the scale of the hip joint. When the peak kinematics of the hip occurred, the peak value of the maximum compression strain was 0.12 (shown in cross-sectional view).

Fig. 1

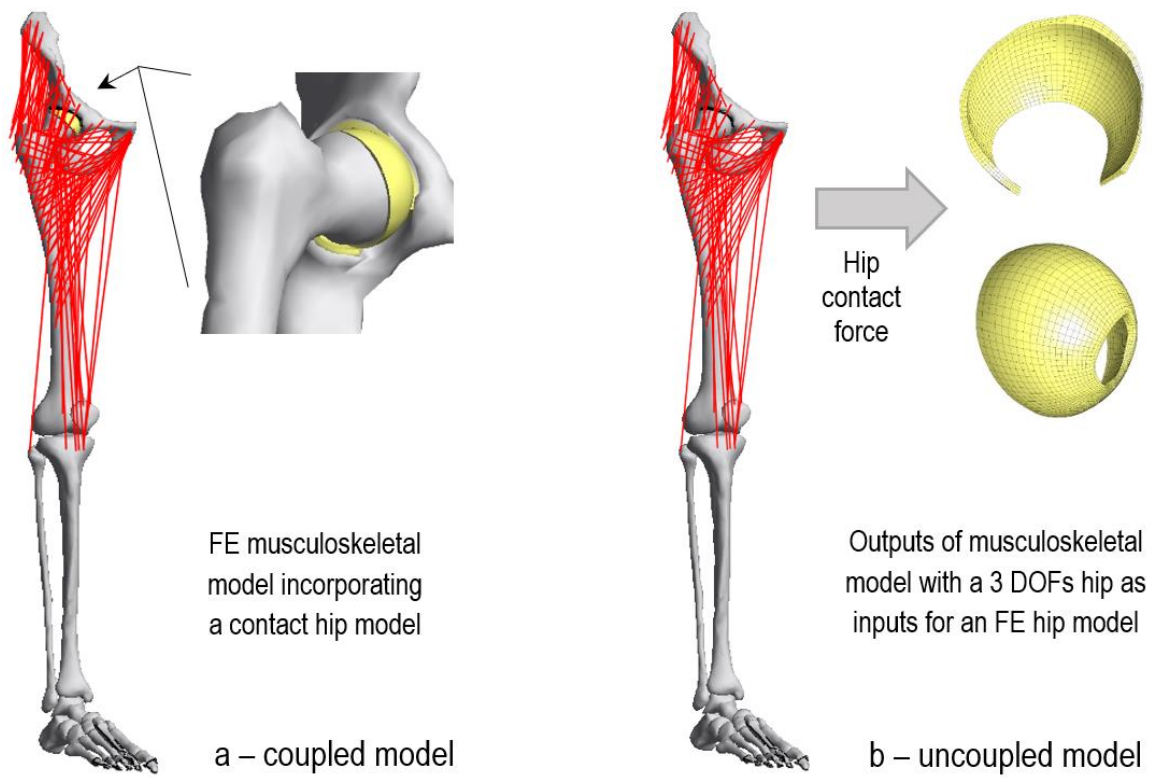


Fig. 2

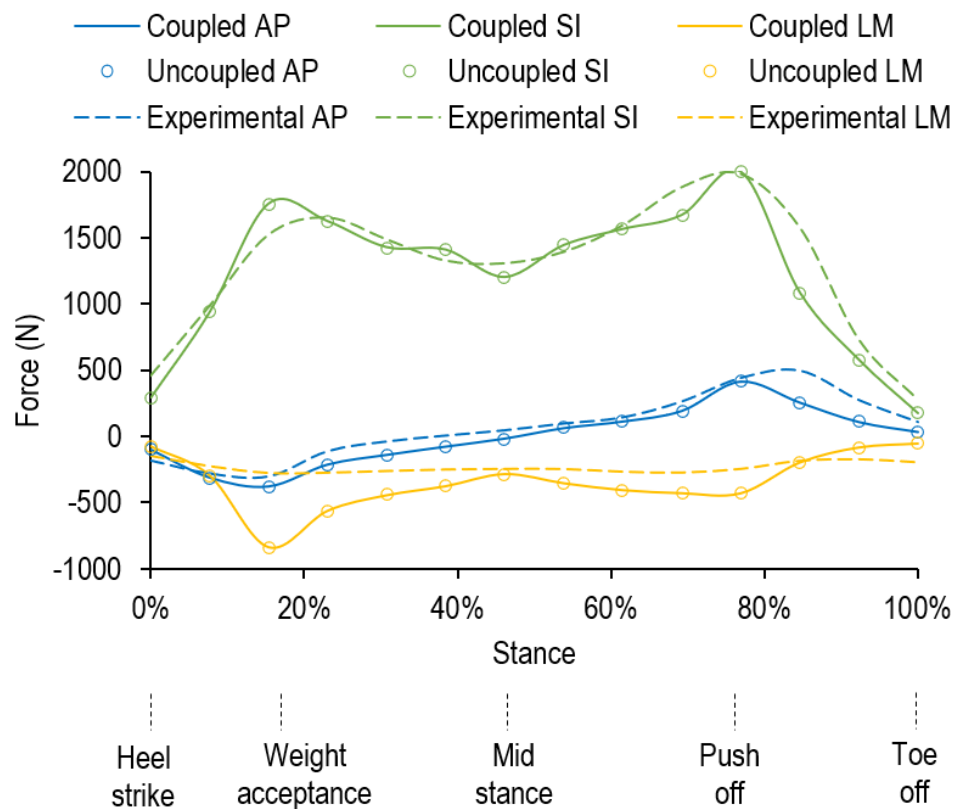


Fig. 3

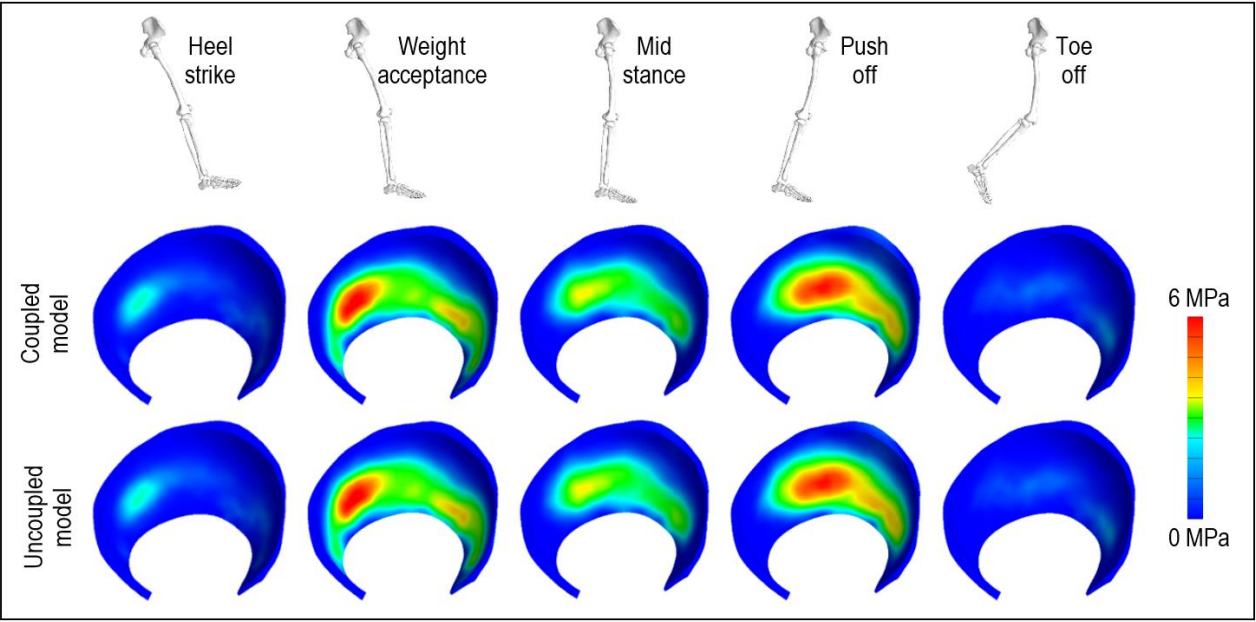


Fig. 4

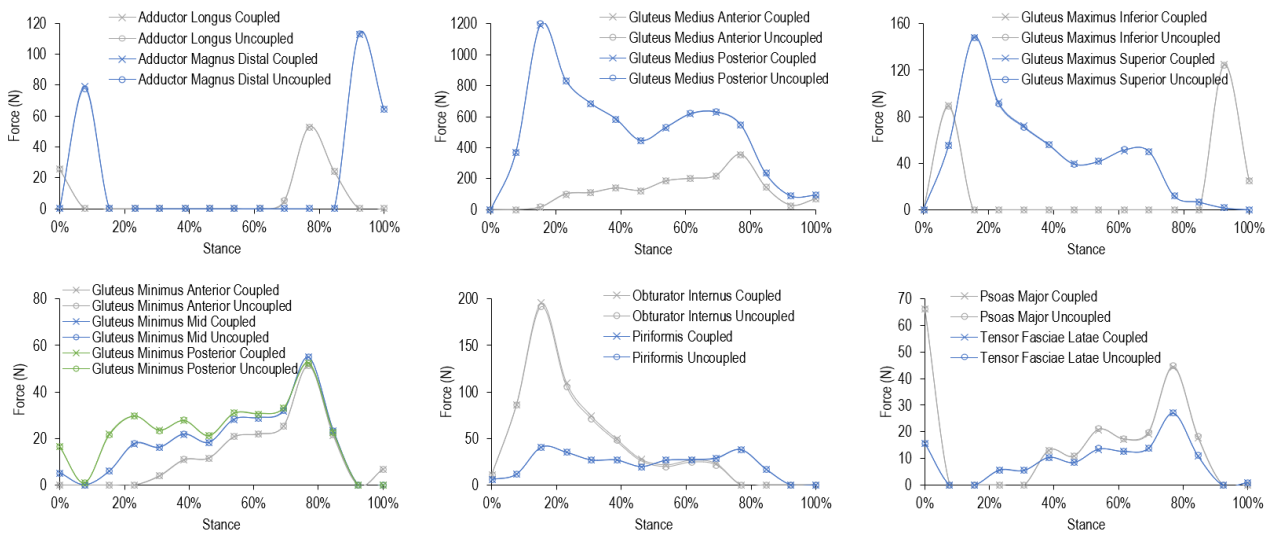




Fig. 5

